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NONINVASIVE HEART RATE MONITOR

FINAL REPORT, PHASE I

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INDUSTRIAL & BIOMEDICAL SENSORS CORPORATION

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The feasibility of obtaining noninvasive heart rate measurements on a chemical warfare garbed subject has been investigated. The device is a portable ultrasonic transceiver which utilizes the Doppler effect to detect movements of the heart wall and blood mass. The periodicity of the heart event is then used to deduce the heart rate measurement. The ultrasonic transmitter and receiver are composite piezoelectric crystals with a reversed-horn section for more efficient energy coupling. The process of imparting and recovering

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sufficient ultrasound energy through the protective clothing is accomplished without using any wetting agent or acoustic gel. In addition to ultrasonic wave detection, the feasibility of passive recording of the hydrostatic cardiac shock wave has also been studied. The low frequency, short duration vibration of the beating heart can be detected using the same piezoelectric receiver. Heart rate monitors based on these two methods have been constructed. Test results show that heart rate can, indeed, be measured even when the subject is wearing a shirt and a sweater, in addition to the standard chemical warfare garment and wrap. However, both monitors failed to give a reliable or usable heart rate measurement when tested on a moving M113 field ambulance. In spite of this limitation, the monitor would still be very useful for quick assessment of the heart condition especially in situations where the casualty is in a protective wrap which precludes the application of a blood pressure cuff.

SUMMARY

The feasibility of obtaining noninvasive heart rate measurements on a chemical warfare garbed subject has been investigated. The device is a portable ultrasonic transceiver which utilizes the Doppler effect to detect movements of the heart wall and blood mass. The periodicity of the heart event is then used to deduce the heart rate measurement. The ultrasonic transmitter and receiver are composite piezoelectric crystals with a reversed-horn section for more efficient energy coupling. The process of imparting and recovering sufficient ultrasound energy through the protective clothing is accomplished without using any wetting agent or acoustic gel. In addition to ultrasonic wave detection, the feasibility of passive recording of the hydrostatic cardiac shock wave has also been studied. The low frequency, short duration vibration of the heart can be detected using the same piezoelectric receiver. Heart rate monitors based on these two methods have been constructed. Test results show that heart rate can, indeed, be measured even when the subject is wearing a shirt and a sweater, in addition to the standard chemical warfare garment and wrap. However, both monitors failed to give a reliable or usable heart rate measurement when tested on a moving M113 field ambulance. In spite of this limitation, the monitor would still be very useful for quick assessment of the heart condition especially in situations where the casualty is in a protective wrap which precludes the application of a blood pressure cuff.

TABLE OF CONTENTS

1.0 INTRODUCTION

This FINAL REPORT is a summary of a Phase I program to determine the feasibility of a noninvasive heart rate monitor for chemical warfare (CW) garbed casualties.

The event of a heart beat can be detected by many different methods. Most of these methods do not lead themselves to the present application either because direct body contact is required or because it is motion artifact sensitive. A brief discussion of various common techniques are given in this section.

1.1 Photoelectric Plethysmograph

Photoelectric plethysmography can be applied to various parts of the body such as finger, toe, ear, forehead, neck, wrist and so on. The choice of location is predicated upon signal-to-noise ratio, ease of application, and signal preservation in clinical shock condition. The light source and photo detector used can be in the visible or infrared range, either narrow or broad band.

Because optical contact with the skin is required, use of this method entails either design modification of the CW suit or provision of a separate internal electronic package in each and every suit. It is conceivable that power can be delivered to the internal package through inductive coupling and measured signal can be telemetered out in a number of ways. However, the added cost may become prohibitive.

Photoelectric finger sensors have been widely used in ambulances where intermediate vibration levels are encountered. In general, tighter compression between the optical sensor and the body will reduce motion artifact but unfortunately it will also reduce blood flow and therefore, the signal strength. The successful adaptation of the photoelectric plethysmograph technique to a high noise environment is still in question.

1.2 Palpatory Method

The heart pulse can be sensed on an artery (radial, carotid, etc.) via a force transducer such as those used in arterial tonometry. Again direct body contact and motion sensitivity render the technique unsuitable for the present application.

A variant of the palpatory method can be used to detect pulse of a CW garbed casualty, externally. In blood pressure measurements involving an occluding cuff, pressure spikes known as oscillometric signals can be detected within the air-filled bladder following each heart beat. These pressure oscillations are generated as the arterial pressure pulse wave interacts with a deformed artery due to the existence of a compression cuff. The pressure spikes can be detected even when the cuff pressure drops to 20-30 mmHg, a pressure which can easily be tolerated. The oscillometric pressure spikes can be detected across an intervening layer of sweater, business suit, ski or leather jacket and the CW protective clothing as well.

The above oscillometric technique can be used to measure heart rate effectively on a CW casualty only in a low noise state. The signal generated by the tiny artery (which is indirectly coupled to the bladder) is small compared to the pressure spikes which may be generated by the arm muscle, vibration and other spurious perturbances. Furthermore, this method is inoperable when the casualty is completely enclosed in a CW body wrap. Variations of the palpatory method such as those involving a small tube of liquid mercury wrapped around the arm are equally ineffective.

1.3 Auscultation Technique

The heart sound can be auscultated directly from the chest or the sternum notch by bearing down on the protective clothing. This approach is less likely to success due to the high noise environment.

1.4 ECG Pickup

Heart rate measurement invloving ECG pickup is by far the most simple and reliable method in a high noise, high vibration environment. The requirement for direct skin contact renders this technique unsuitable in the present application.

Due to the availability of low voltage, low current operational amplifiers, the amplification and telemetering of ECG signals can be achieved with very low power easily obtainable from external source by induction coupling. Therefore, if the installation of an internal

electronic package is allowed, ECG would be the method of choice. The electrodes can be constructed of bare metal (without gel) judiciously located within the CW garment.

1.5 Current Approach

The conventional methods of heart rate measurement have been reviewed and were found to be inadequate for the present application. The method under current consideration involves velocity measurements of the heart wall based on the ultrasound Doppler technique. The possibility of heart rate measurement based on the detection of low frequency, hydrostatic, cardiac vibration wave is also explored.

2.0 METHODOLOGY

The basic methodology and underlying principle of the heart rate monitor under development is discussed in this section.

2.1 Ultrasound Technique

Ultrasound has been widely used in diagnostic medicine for non-invasive imaging of internal organs or structures of the human body. A clear, moving image of a beating heart can be displayed continuously on a monitor. Information regarding heart rate can be extracted in a number of ways. In the present application, the Doppler effect will be utilized.

The principle of operation is illustrated in Figure 1. A continuous ultrasonic wave is transmitted to the heart while the returning echoes are picked up by a separate receiving transducer. As the heart moves, the frequency of the echo changes due to the Doppler effect. The amount of frequency shift Δf can be expressed in the following formula:

$$\Delta f = \frac{2V}{\lambda} = 2f \frac{V}{C} \qquad (1)$$

where V = velocity of heart wall

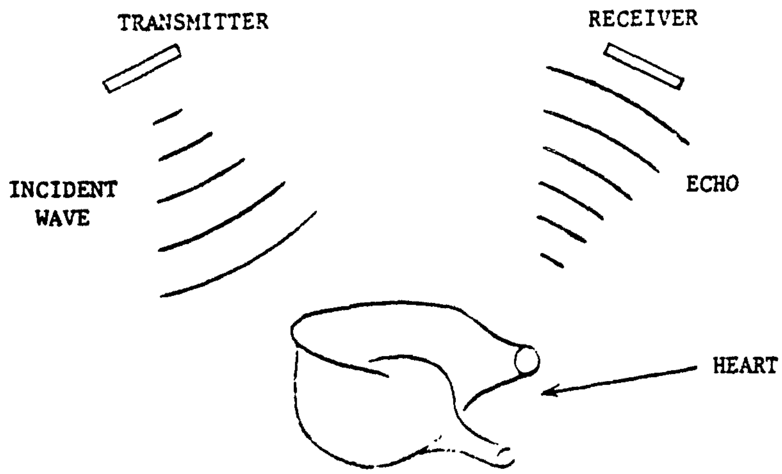
λ = wave length of the transmitted ultrasound

f = frequency of incident wave

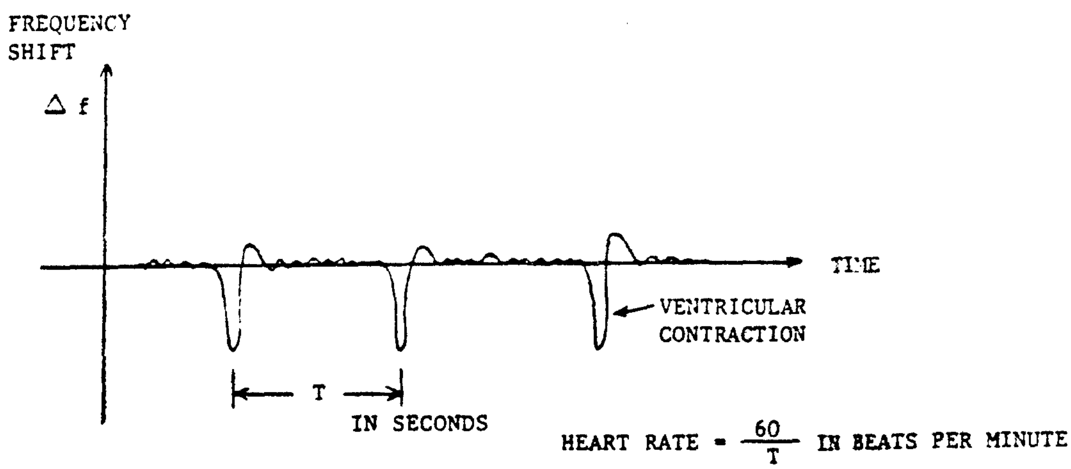
C = speed of sound (1.55×10^5 cm/sec)

The beat frequency of the incident ultrasound and the echo (i.e., Δf) can be detected and converted to a time varying voltage (Figure 1-b) from which the heart rate can be deduced. The maximum frequency shift is expected to happen during ventricular contraction, and the magnitude is estimated to be on the order of 0.013%, assuming the maximum velocity of the heart wall is on the order of 10 cm/sec.

With a suitable design, sufficient energy can be coupled to the heart and reflected from the heart wall to the receiver outside the



(a) WAVE PATH OF ULTRASOUND



(b) SCHEMATIC REPRESENTATION OF THE FREQUENCY SHIFT

FIGURE 1 HEART RATE MEASUREMENT BASED ON THE ULTRASOUND DOPPLER TECHNIQUE

protective CW garment.

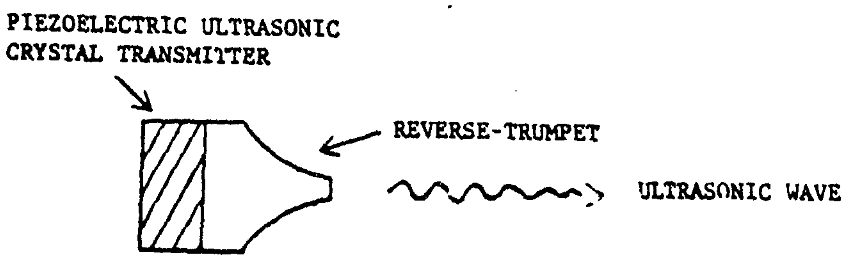
2.1.1 Energy Coupling

Whenever ultrasound passes from one medium to another, a portion of the energy is reflected and the remainder is refracted. The amount of usable energy refracted depends on the difference in acoustic impedance (defined as the product of speed of sound and the density) and the incident angle. At normal incidence (i.e., perpendicular incidence) and at interfaces of extreme difference in media such as between air and tissue, almost all the energy will be reflected and practically none will continue through the tissue. Normally, in applying ultrasound to the body, an airless contact is produced through the use of a gel, which is precluded in the present application.

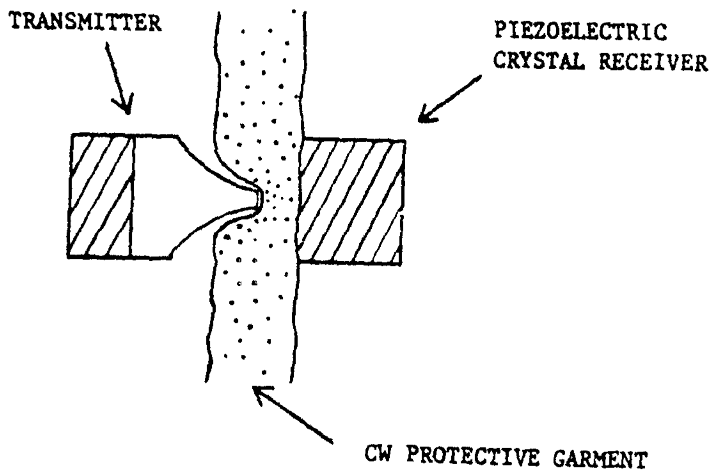
Furthermore, in the present application, there is an intervening layer of CW protective clothing between the transceiver device and the body. This layer acts as a serious barrier for ultrasound energy. In order to overcome this difficulty, several plausible approaches have been investigated. These included special transmitters with a oil-filled compartment, with a silicone rubber interface, and with an epoxy filled, impedance matching layer. None of these methods have resulted in any significant ultrasound penetration through the CW garment material.

However, insights gained in these preliminary studies have resulted in a new approach to the problem. It was discovered that if a nose cone section in the form of a reversed trumpet or horn, is installed on the transmitter tip (Figure 2), the received signal is greatly enhanced. Because of the area transformation, the maximum energy density at the tip of the transmitter can be magnified by the area ratio. In addition, the contact pressure at the tip is also increased by the same area ratio. Therefore, the layer of CW garment underneath the tip is more compacted, contain less air, and become less lossy. These two factors are believed to be responsible for the great enhancement in energy transmission.

The enhancement effect due to the reverse-trumpet design was



(a) ULTRASONIC TRANSMITTER WITH A REVERSE-TRUMPET NOSE SECTION



(b) APPLICATION OF THE REVERSE-TRUMPET TRANSMITTER

FIGURE 2 SCHEMATIC DRAWING OF A REVERSE-TRUMPET TRANSMITTER

demonstrated by a simple transmission comparison test (Figure 3). Identical pairs of ultrasonic transmitters and receivers were mounted across two layers of CW protective garment (8415-00-407-1062) or body wrap (8465-01-079-9875) under a constant holding force equal to 0.3 Kg. The transmitters were driven by a sine wave generator at their respective optimum frequency (i.e., a frequency at which a maximum output is received). The ultrasonic wave reaching the receiver was measured directly by an oscilloscope with 10M ohm input impedance. In this simple test, two layers of the CW protective material were used in order to simulate the actual heart rate measurement configuration.

Four different ultrasonic transmitter and receiver pairs were used in the transmission test. These are:

Pair A

Frequency: 210 KHz

Diameter: 1.3 cm

Model Number: E - 188

Manufacturer: Massa Products Corporation, Hingham, MA

Description: Transmitting Sensitivity = + 20 dB vs 1 μbar/ft/volt

Receiving Sensitivity = - 77 dB vs 1 volt/ μbar

Total Beam Angle at - 3 dB = 10^{o}

Surface Protective Material: Silicone Rubber

Pair B

Frequency: 200 KHz

Diameter: 3.8 cm

Model Number: R - 283 E

Manufacturer: Massa Products Corporation, Hingham, MA

Description: Transmitting Sensitivity = + 91.5 dB vs 1 μbar/yard/watt

Receiving Sensitivity = - 76.5 dB vs 1 volt/ μbar

Total Beam Angle at - 3 dB = 13^{o}

Surface Protective Material: Epoxy

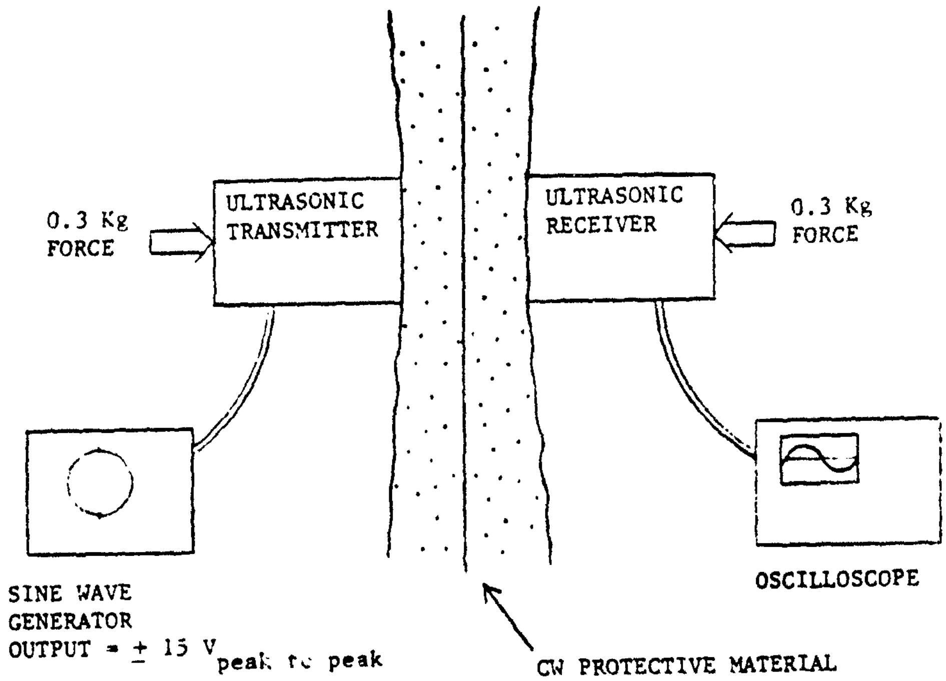


FIGURE 3 SETUP FOR A SIMPLE TRANSMISSION TEST

Pair C

Frequency: 210 KHz

Diameter: 1.0 cm

Model Number: 2-971-05360-005

Manufacturer: Ferroxcube Corporation, Saugerties, NY

Description: Thickness: 0.3 cm

Material: PXE5, Lead Zirconate/Lead Titanate

Pair D

Frequency: 205 KHz

Diameter: Transmitter: 1.0 cm

Aluminum Horn (Tip): 0.3 cm

Receiver: 1.0 cm

Model Number: 2-971-05360-005

Manufacturer: Ferroxcube Corporation, Saugerties, NY

Description: Aluminum horn was attached to the transmitting crystal by EASTMAN 910 glue

Results of the transmission test are summarized in Table 1 below. In addition to the data presented for the CW garment and body wrap material, those for a 3 mm thick, homogeneous rubber material are also included for comparison. It is quite obvious that the energy transmitted is greatly enhanced when a reversed-trumpet, nose section is added to the transmitting crystal.

Therefore, the nose section of the ultrasonic transducer becomes a key element in the current development. The design of an optimum reversed-trumpet section is presented in the next section.

TABLE 1

A Comparison of Transmitted Signal (in Volts peak to peak) Across Rubber and CW Protective Material Layers

	Rubber V_{p-p}	CW Garment Material V_{p-p}	CW Body Wrap Material V_{p-p}
Pair A	0.9	0	0
Pair B	1.5	0.01	0.015
Pair C	0.8	0.01	0.015
Pair D	5.5	1.5	1.9

2.1.2 Transmitter Design

Ultrasounds can be generated by exciting piezoelectric crystal oscillators. The design of these oscillators has been described in numerous texts and background information relevent to the present application can be found in Ref. 1 through 6.

One of the optimum transmitter designs for the present application is schematically represented in Figure 4. The transmitter is composed of a half-wavelength oscillator section and a half-wavelength reverse-trumpet section. Based on simple transmission line analysis of the Helmholtz equation, the maximum attainable exit velocity can be obtained only with a stepwise varying reverse-trumpet having a quarter-wavelength base section and tip section as shown in Figure 4. Because the trumpet section is made of a very strong and resilent material (G-10 epoxy), and because the ultrasonic energy required is relatively small, the maximum stress exists at the junction of section 4 and 5 (Figure 4) can well be tolerated.

The half-wavelength oscillator is a composite transducer consisting of three portions, a high acoustic impedance end mass, a piezoelectric ceramic crystal middle portion, and a low acoustic impedance piston. The heavy end mass is used to reduce the energy loss in the inactive direction. The light piston is used to amplify the velocity and therefore the wave energy. The piezoelectric crystal is polarized in the x-direction and it is operated in the thickness mode.

Based on tabulated material constants and chosen dimensions, the frequency and various characteristics of the composite oscillator can be calculated by the standard transmission line analysis. Therefore, through iteration, any oscillator with a specified frequency specification can be designed and constructed. The dimensions of the reverse-trumpet section can be separately calculated, assuming the addition of this section does not change the final frequency to any significant degree. Otherwise, more iterations are required if the reverse-trumpet section is incorporated in the transmission line

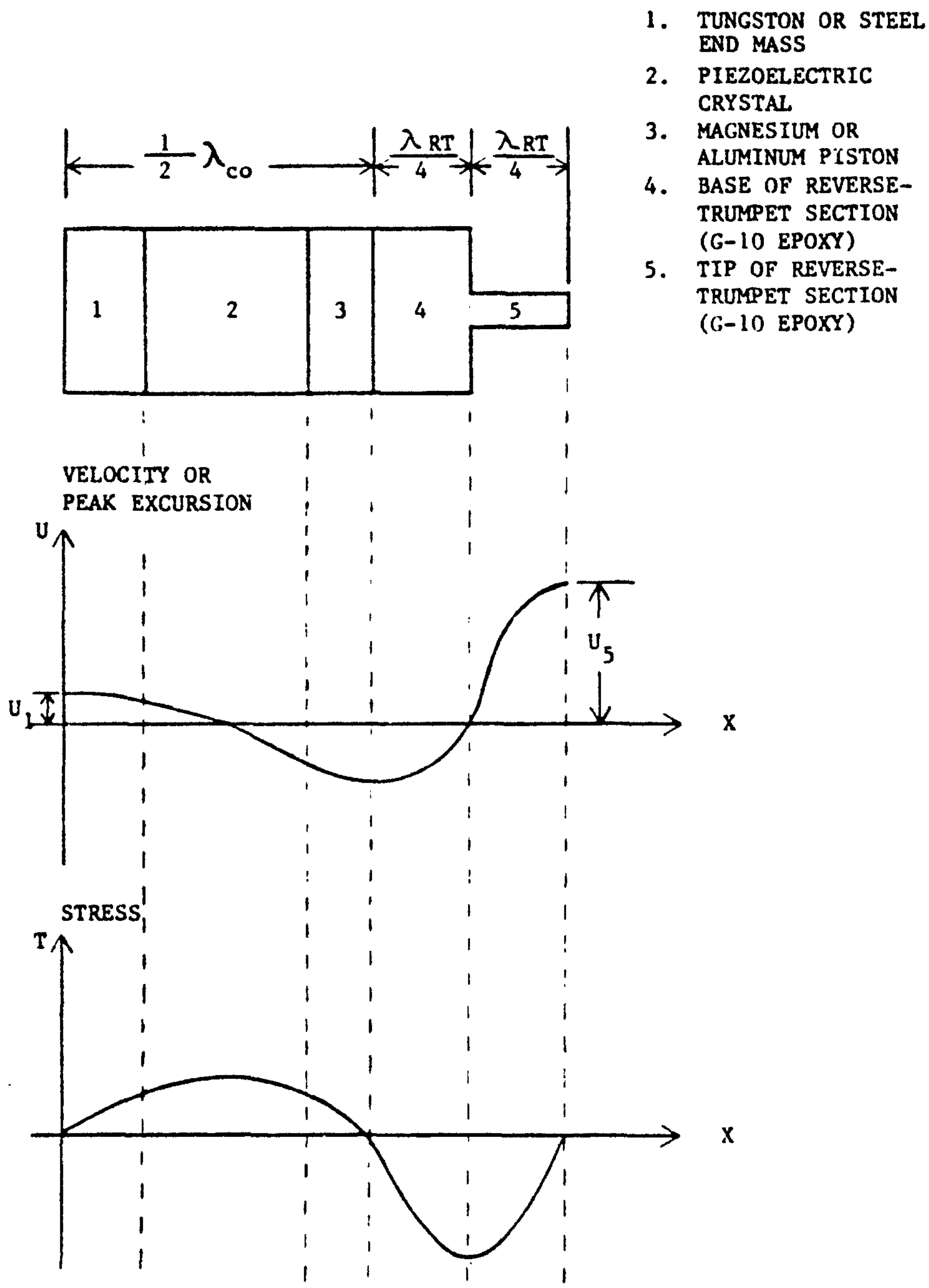


FIGURE 4 TRANSMITTER DESIGN BASED ON THE THICKNESS MODE

calculation for a full-wavelength composite transmitter.

The transmitter shown in Figure 4 is designed to operate in the thickness mode and there is no theoretical limitation on the shape of the cross-section. As borne out by actual fabrication and testing, this type of transmitter is most efficient and has high radiation intensity. However, its use in the present application is not favored due to the following shortcomings:

1. The overall length of the transmitter tends to be too long to satisfy the low profile requirement of the portable heart rate monitor.
2. The length of the piezoelectric crystal portion is such that a high excitation voltage is required. This condition is incompatible with a battery-operated monitor. Although the crystal portion can be made up of short segments biased in a parallel fashion in order to reduce voltage drive, the design and construction of this segmented oscillator may become prohibitively difficult.
3. In practice, it is sometimes difficult to satisfy the condition that the thickness of the piezoelectric crystal be larger or at least comparable to the diameter or lateral dimension. As a result, a whole series of anharmonic spurious resonances may be produced. This results in the resonance being rather complex, with cross coupling between the radial and thickness directions. These unwanted resonances may occur in the required frequency range of the basic thickness mode. Only with some difficulty can these spurious resonances be suppressed by varying the electrode covering area.

In view of these pitfalls, the thickness mode transmitter is not used in favor of a transmitter operating in the fundamental radial mode(Figure 5). This transmitter is composed of a piezoelectric crystal portion and a reverse-trumpet portion. The function and design of the trumpet portion remains the same as before for the

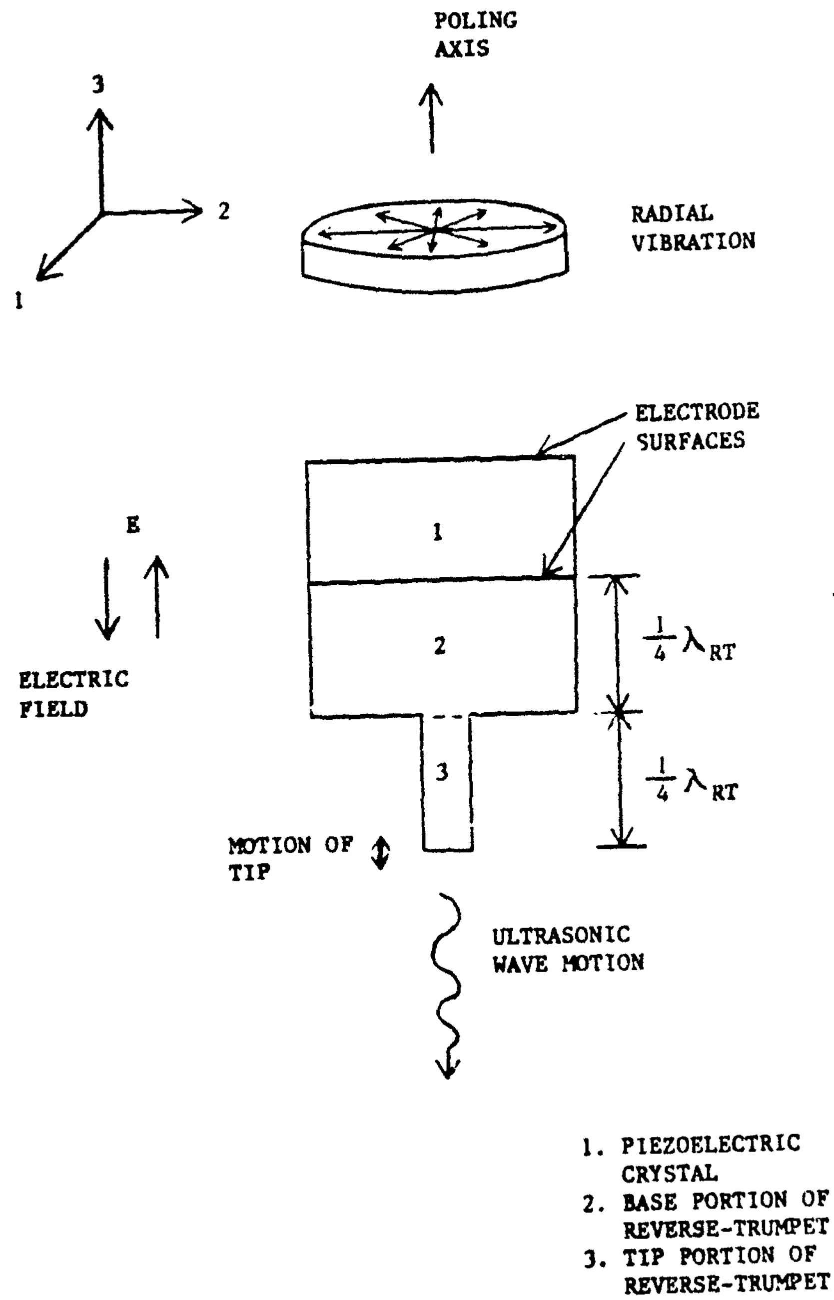


FIGURE 5 SCHEMATIC DRAWING OF A TRANSMITTER BASED ON THE RADIAL MODE

thickness mode oscillator. However, the piezoelectric crystal is now oscillating in the radial direction even though the electric field is still applied in the axial and poling direction (Figure 5). The radial mechanical motion induces expansion and contraction in the axial direction, producing an axial, ultrasonic vibration directly coupled to the reverse-trumpet section. The induced axial vibration is governed by the THICKNESS COUPLING COEFFICIENT commonly denoted as K_t which represents the coupling between the electric field in the axial direction (thickness direction) and the mechanical vibration in the axial direction. K_t is in general smaller than K_{33} (coupling coefficient for the thickness mode) because of the constraint imposed by the large lateral dimensions of the ceramic crystal disc relative to the thickness.

The advantages of the transmitter based on the radial mode are as follows:

1. low profile construction
2. thin disc structure which is more amenable to low voltage excitation
3. the radial mode resonance is always at a far lower frequency than thickness or other regular or spurious anharmonic modes.
4. ease of design as a result of the direct relationship between frequency and diameter

Transmitters based on the radial mode have been constructed for heart rate measurement. The piezoelectric ceramic crystal used has the following relevant properties:

Trade Name: PXE5
Manufacturer: Ferroxcube Corporation
Saugerties, NY
Material: Lead zirconate/lead titanate
Frequency Constants: radial mode 2000 Hz-m
thickness mode 1900 Hz-m
Density: 7.65×10^3 Kg/m^3

Poisson's Ratio: 0.3

Mechanical Quality Factor for Radial Mode = 80

Coupling Coefficient: radial Kp = 0.6

thickness K_{33} = 0.69

Voltage Constant: thickness $g_{33} = 22.7 \times 10^{-3}$ Vm/N

The diameters of the ceramic crystal used included 1.0 cm, 1.6 cm and 2.54 cm whereas the thickness varies from 0.1 to 0.5 cm. The diameter of the reverse-trumpet is 0.3 cm at the tip for all crystal diameters. The crystal oscillator in the radial mode can also be composed of thin discs connected in a parallel fashion in order to enhance the applied electric field strength. The multiple ceramic crystal discs were bonded together by high strength epoxy cured at an elavated temperature under a moderate clamping pressure. Electrical connections were provided by fine silver screens sandwiched between two adjacent crystals. The epoxy and screen were obtained from the following sources:

Epoxy

Trade Name: ECCOBOND 104

Manufacturer: Emerson & Cuming
Canton, Massachusetts

Screen

Part Number: 5 Ag 7-4/0

Manufacturer: Exmet Corporation
Bridgeport, Connecticut

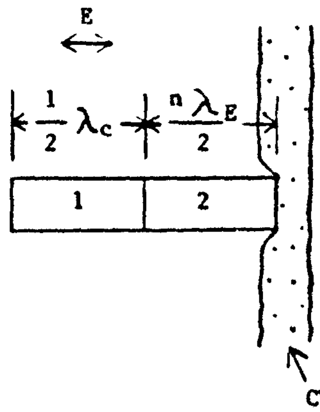
2.1.3 Receiver Design

The design for an ultrasonic receiver for the present application is simpler and less involved than that for the transmitter. The receiver must have an appropriate resonant frequency coinciding with that of the transmitter. Also the receiver must have a slender tip resembling the tip section of the reverse-trumpet in order to maintain a high contact pressure on the CW protective layer.

A receiver design based on the thickness mode is schematically represented in Figure 6-a. The extension rod must have a small diameter (0.3 cm) and a length equal to any integral multiples of the half-wavelength pertaining to the material (G-10 epoxy, for instance). The piezoelectric crystal oscillator is a half-wavelength resonator operating in the thickness mode. The cross sectional area of the crystal need not be identical to the extension rod.

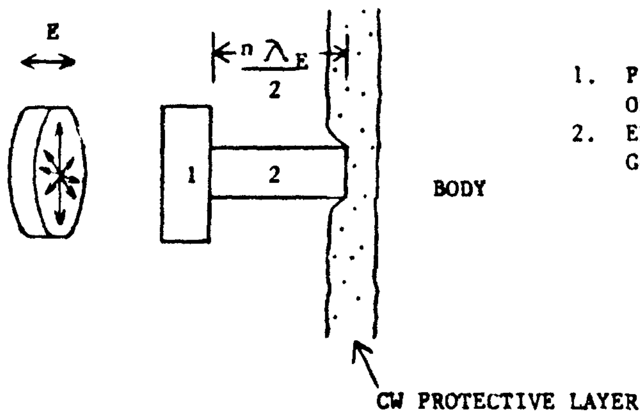
A receiver based on the radial mode excitation is shown in Figure 6-b. The extension rod transmits the axial ultrasonic motion of repeated expansion and compression to the piezoelectric disc, thereby exciting the resonance in the radial direction. The resulting voltage signal output is proportional to the thickness of the ceramic disc.

The resonant frequencies of transmitters and receivers can be measured by a simple impedometer device. A typical curve of transducer impedance versus frequency such as the one shown in Figure 7 can be obtained. The transmitter is normally driven at or near the resonance frequency for maximum efficiency and ultrasound output. However, in the case of a receiver, the oscillator is so designed that its anti-resonance frequency coincides with the resonant frequency of the transmitter for maximum sensitivity.



1. PIEZOELECTRIC CRYSTAL OSCILLATOR
2. EXTENSION ROD MADE OF G-10 EPOXY MATERIAL

(a) ULTRASONIC RECEIVER BASED ON THE THICKNESS MODE EXCITATION



1. PIEZOELECTRIC DISC OSCILLATOR
2. EXTENSION ROD MADE OF G-10 EPOXY MATERIAL

(b) ULTRASONIC RECEIVER BASED ON RADIAL MODE EXCITATION

FIGURE 6 SCHEMATIC DESIGN OF ULTRASONIC RECEIVERS

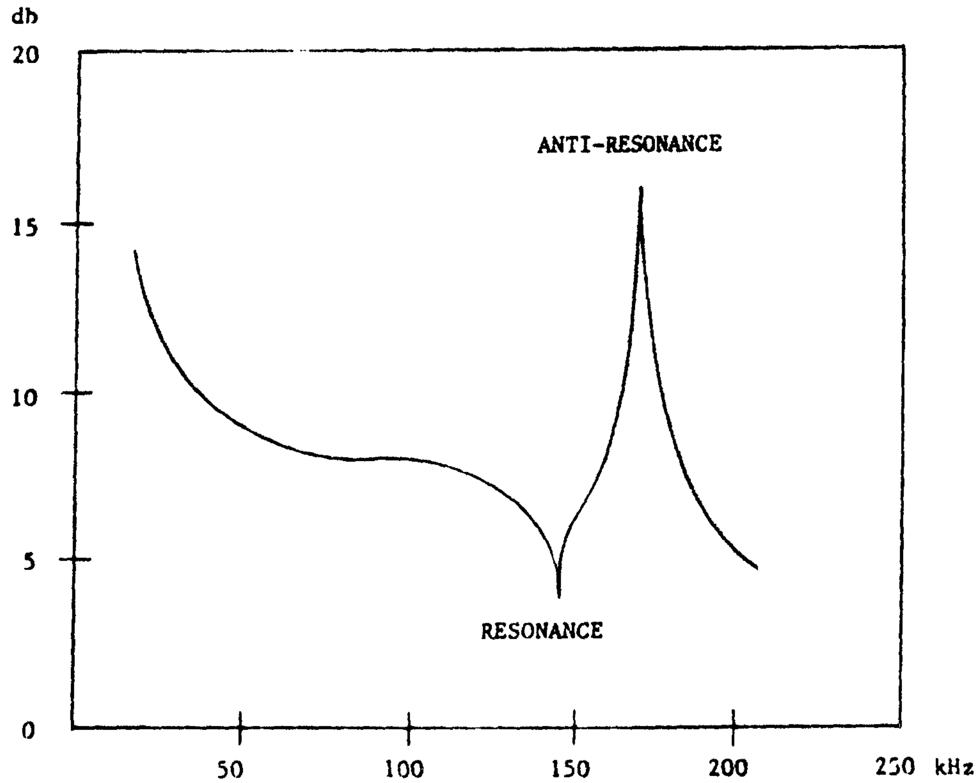


FIGURE 7 IMPEDANCE OF TRANSDUCER VERSUS FREQUENCY

2.1.4 Operating Frequency Range

In ultrasound velocity measrurements, a higher frequency for the carrier wave is normally preferred because it will result in a more pronounced Doppler effect, a larger frequency shift and therefore a better signal-to-noise ratio. Unfortunately, the absorption of sound energy by body tissues is frequency dependant and high frequency wave generally has a very shallow penetration depth. In order to detect an echo bouncing off the heart wall, the maximum practical frequency is limited to below 1 MHz.

In the present application, the lossy, CW protective garments would impose further limitations on the frequency. The transmission of ultrasound at different frequencies has been measured, using the setup of Figure 3. The results are summarized in Table 2 below.

TABLE 2

Transmission of Ultrasound at Different Frequencies

	Rubber	CW Garment Material	CW Body Wrap Material
	V_{p-p}	V_{p-p}	V_{p-p}
199 kHz	12	8	9
300 kHz	1.2	0.15	0.25
450 kHz	0.3	0.04	0.04
1100 kHz	0.1	0.005	0.01

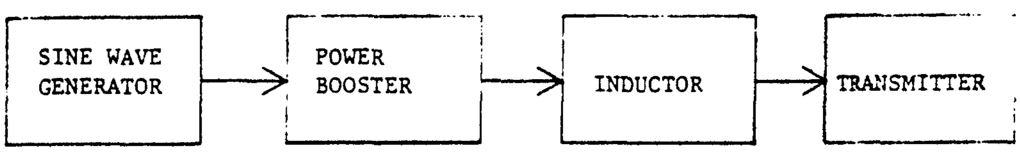
The transducers used in the test were designed with a reverse-trumpet nose section. Results of Table 2 indicated that operation of the transducers at higher frequencies is not desirable due to higher absorption, less efficient transmitter and less sensitive receiver. It therefore appears that the optimum operating frequency range is around 200 kHz.

Finally, it should be mentioned that even in the absence of absorption, the transmission of sound energy is frequency dependent based on transmission line analysis. To demonstrate

this effect, a simple computer simulation/calculation was performed on the transmission loss of a layer of sheepskin (Figure 8). For a thickness equal to 0.3 cm, the ultrasound can pass through the sheepskin layer with no loss at frequencies near 75, 150, 225, ... and so on when the thickness of the sheepskin layer is equal to multiples of half-wavelength. The acoustic properties of sheepskin is tabulated in Ref. 2. Sheepskin is the only material which has some resemblance to the CW garment and wrap material and for which the acoustic properties are published.

2.1.5 Excitation of Transmitter and Receiver

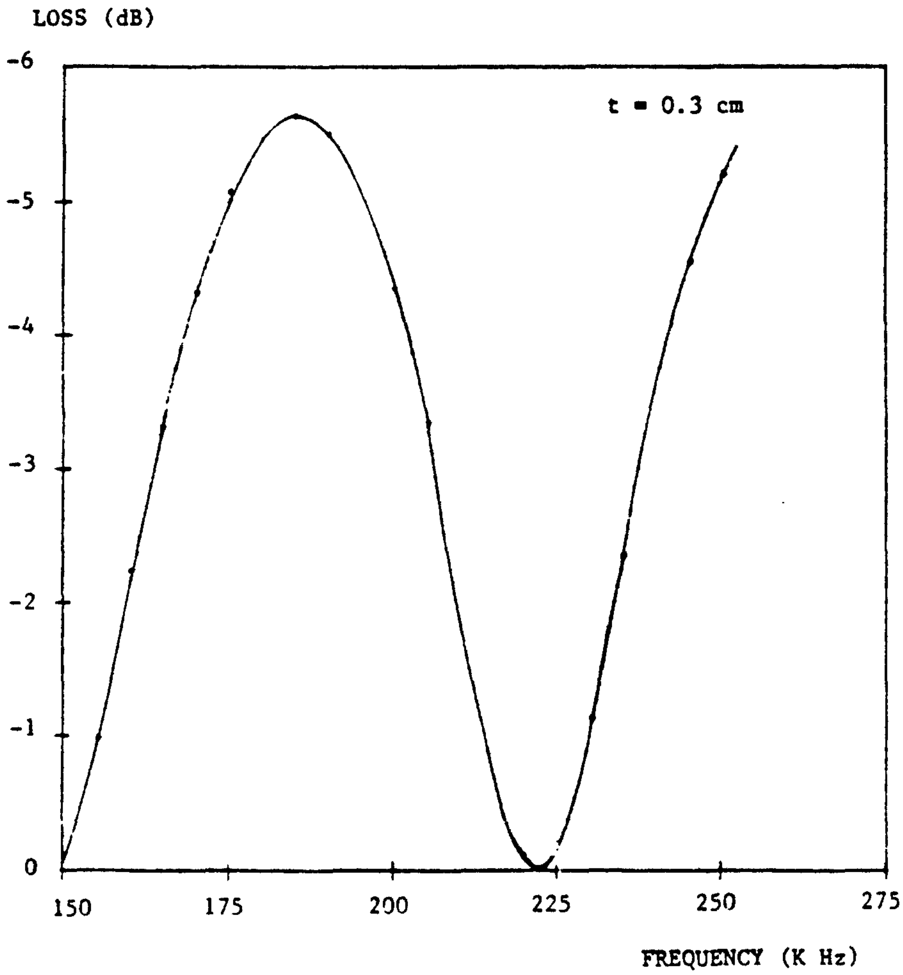
In the present application, the transmitter can be excited by a resonant circuit or it can be driven by a constant frequency source. The latter approach is normally more straight forward and uncomplicated. The functional circuit elements involved are as follows:

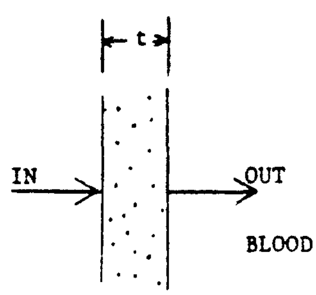


The sine wave generator and the power booster are small commercial integrated circuits and the inductor, the low resistance type, is used to series-tune the transmitter.

The circuitry required for processing the received signals are more complicated. Due to the severe attenuation of the ultrasound at high frequencies, the operating frequency was chosen to be approximately 200 KHz. At this relatively low frequency, the Doppler effect is small and the beat frequency is low. Special circuitry needed to process these signals is described in this section.

The functional block diagram for detection of heart beat is shown in Figure 9. The received echo signal is first heterodyned with the transmitter frequency via the mixer stage. The low-pass





$$\text{LOSS} = 10 \text{ LOG} \left[\frac{\text{INTENSITY(OUT)}}{\text{INTENSITY(IN)}} \right]$$

FOR SHEEPSKIN

$C = 470$ m/sec

$\rho C = 4.23 \times 10^5$ Kg/sec/m^2

FOR BLOOD

$C = 1550$ m/sec

$\rho C = 1.56 \times 10^6$ Kg/sec/m^2

FIGURE 8 TRANSMISSION LOSS ACROSS A SHEEPSKIN LAYER

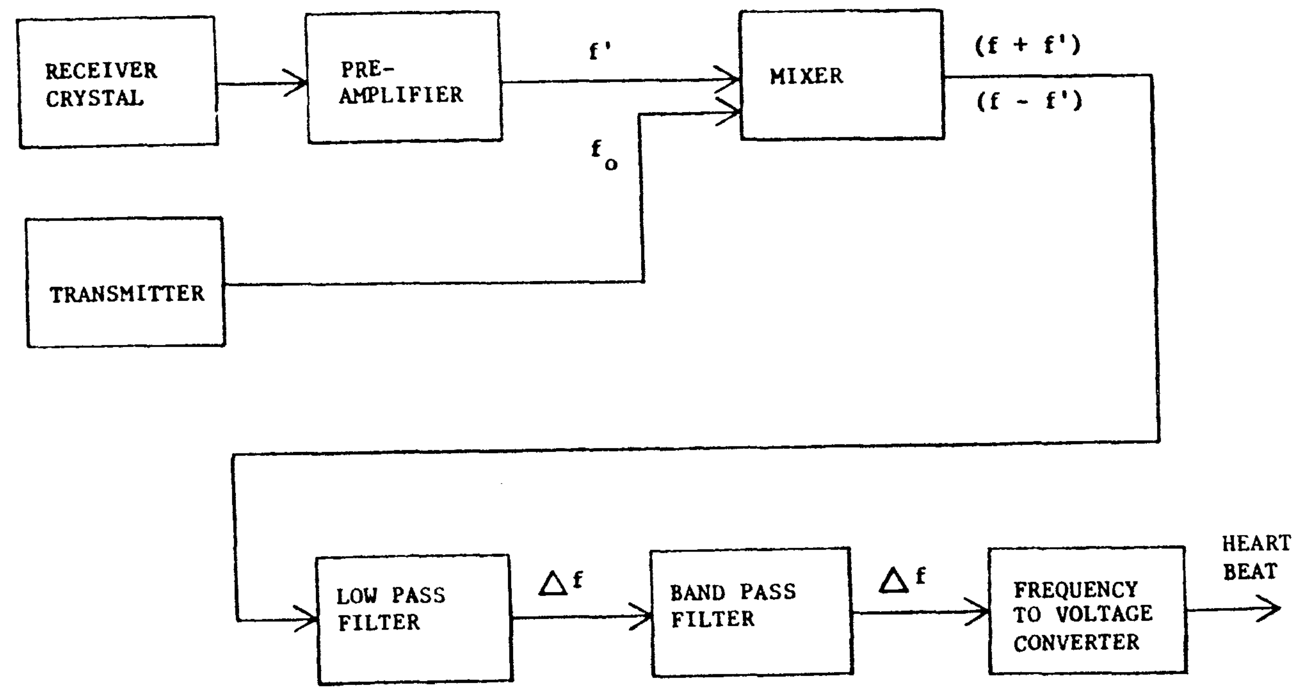


FIGURE 9 FUNCTIONAL BLOCK DIAGRAM OF SIGNAL PROCESSING CIRCUITRY TO DETECT HEART BEAT

filter is used to reject all frequencies higher than the beat frequency. A band-pass filter is also added to better select the narrow pass band for the beat frequency. The filtered and processed beat frequency is then converted to a voltage signal by means of the F-to-V converter.

The circuit elements needed for each functional block are, in general, both simple and standard in nature. A detailed description can be omitted here for brevity. A typical recording of the processed beat frequency signal is shown in Figure 10.

2.1.6 Transmitter-Receiver Array

Transmitters and receivers were arranged in a cluster to form an array. Multiple transmitters were used to ensure a more even distribution of the ultrasonic wave energy, whereas multiple receivers were used to lessen the effect of imprecise placement. In the final design, the array consisted of two identical transmitters and two receivers arranged symmetrically and alternatingly on a circle. The transmitters were excited in a synchronous fashion with identical frequency and phase. The signals from the receivers were processed separately and the presence of a substantial beat frequency output from any channel is used as the criterium for heart beat detection.

The transmitter-receiver arrays were housed in metal casings with the G-10 epoxy trumpets protruding from the opening. It was found necessary to lengthen the tip portion of the reverse-trumpet to three-quarter wavelength. Different methods of attaching the individual crystal oscillator to the casing have been investigated. Test results indicated that the receiver oscillator can be cemented to the bottom of the casing by means of high strength epoxy. The transmitter crystal can be attached to the casing via a thin, intervening layer of cork material. The shoulder of each trumpet was isolated by a thin layer of foam material. The cavity inside the metal housing was filled with a non-stick, light weight, silicone rubber resin.

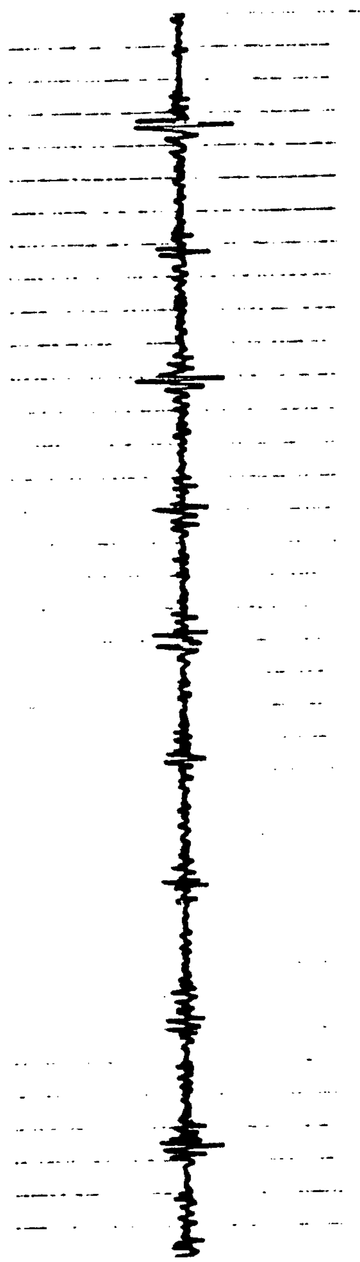


FIGURE 10 A TYPICAL RECORDING OF THE PROCESSED BEAT FREQUENCY

2.2 Hydrostatic Wave Detection

The heart is surrounded by essentially incompressible body fluids and tissues. Because of its relatively violent motion, certain low frequency, hydrostatic wave or vibration is generated which propagates outward to the chest surface at normal cardiac frequency. The resulting tremor on the chest wall is normally visible and it can be detected by the microwave technique. This hydrostatic wave can also be easily measured by means of a piezoelectric transducer.

If the ultrasonic transmitter in Figure 1-a is removed, the remaining piezoelectric receiver will register the signature of a low frequency wave (Figure 11). This hydrostatic wave is in synchronization with the heart movement and can therefore be utilized to measure the heart rate. In this case, the piezoelectric receiver is acting as an accelerometer, registering the "shock" wave traversing the CW garment.

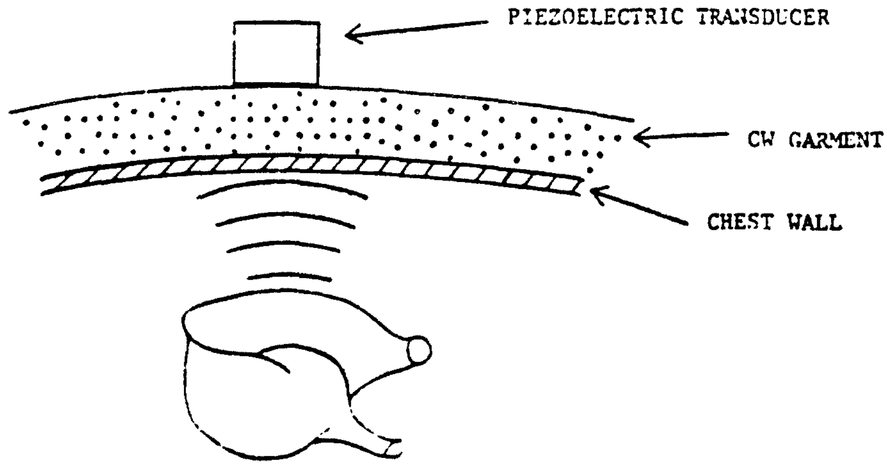
The efficacy of both the ultrasound Doppler method and the hydrostatic wave detection method will be investigated in sections to follow.

2.3 Appearance and Operation of Heart Rate Monitor

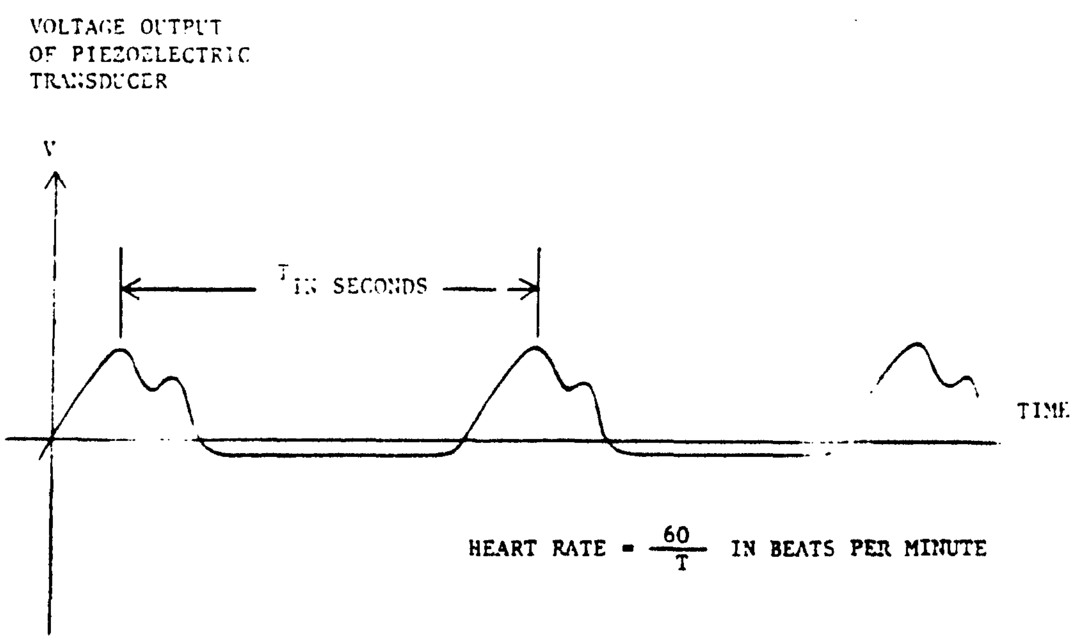
This section describes the projected features, functional appearance, and operation of the heart rate monitor.

The construction and panel layout is illustrated in Figure 12. The heart rate measurement is shown digitally on the top part of the unit. A green, light emitting diode is also illuminated in a synchronous fashion to indicate the cardiac rhythm. Ears extended from the ends of the metal casing are provided to anchor an elastic belt used to strap the unit securely on the chest for continuous heart rate monitoring. The monitor can also be operated in a hand-held fashion to obtain a quick measurement of the heart rate.

Other features incorporated in the design of the monitor are:



(a) HYDROSTATIC, CARDIAC WAVE



(b) SCHEMATIC SIGNATURE OF HYDROSTATIC CARDIAC WAVE

FIGURE 11 HEART RATE MEASUREMENT BASED ON HYDROSTATIC WAVE DETECTION

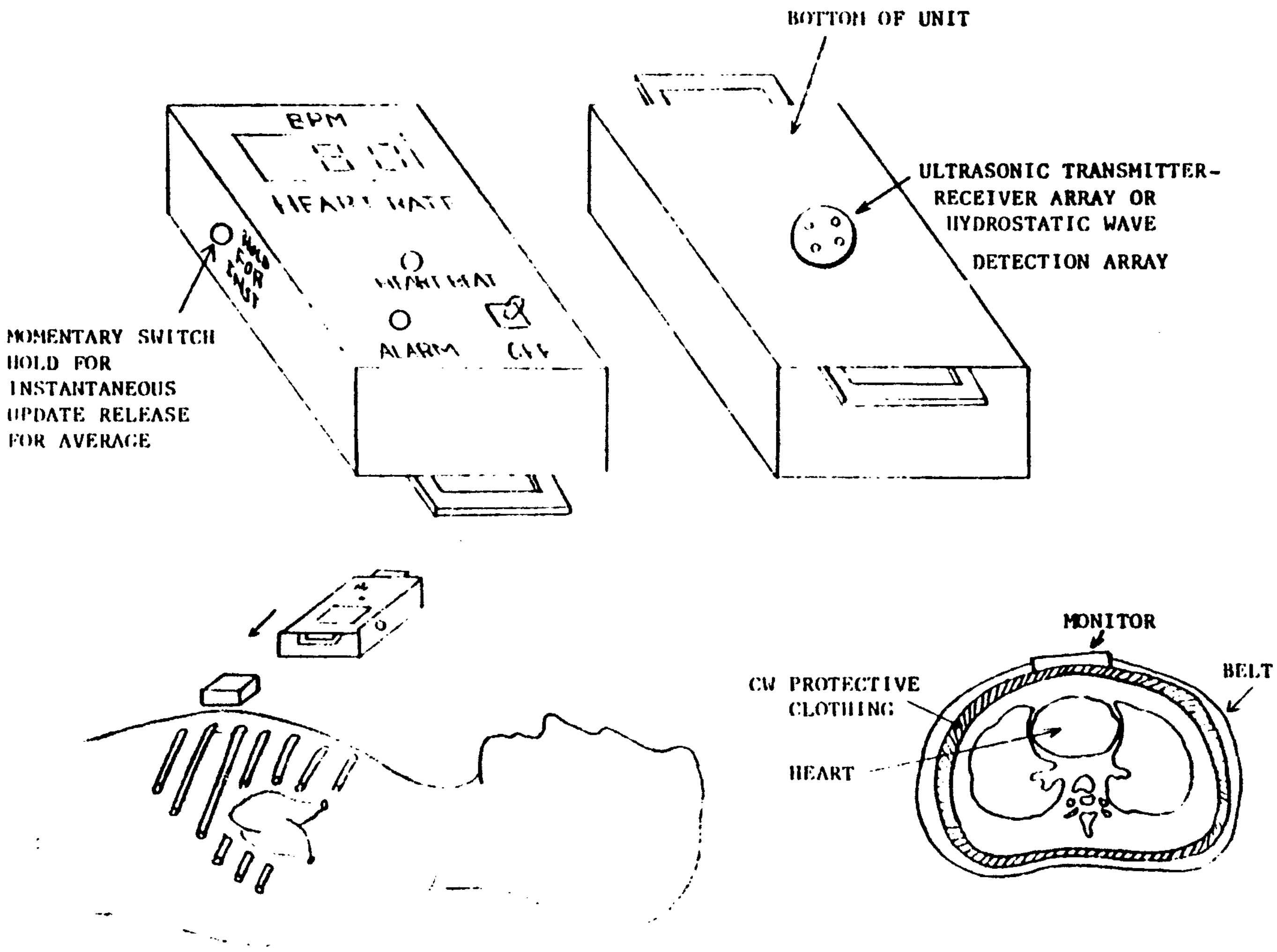


FIGURE 12 PROJECTED FEATURES AND OPERATION OF THE HEART RATE MONITOR

(1) Battery Charging - The battery charging circuit is designed to operate on a 24 VDC external source. When the power source is 110 VAC, an adaptor cable with a small step-down transformer, (similar to those used for hand-held calculator) will be used. On the transformer casing, a small slide switch will be installed which will be used for coil switching when the power source is 220 VAC.

(2) Alarm - An alarm in the form of a flashing red light has been included in the design. This alarm will be actuated whenever the heart rate falls below or exceeds the preset levels. The alarm limits can be altered only through adjustments of internal trimmers located on the circuit board.

(3) Selectable Instantaneous/Average Heart Rate Measurement - The present monitor is designed to measure the time span between consecutive heart beats and converts it to rate information in real time. A "momentary" type of switch will be installed on the side of the monitor. Whenever this switch is depressed, the monitor enters the "INSTANTANEOUS" mode, and the heart rate is presented continuously and it is updated with each new heart beat. This feature is useful for a quick assessment of the heart rate. When the switch is released, the monitor reverts to the normal "AVERAGE" mode in which the heart rate presented is averaged over several (approximately 4) previous beats.

(4) Fast Battery Recharging System - Ni - Cd batteries will be used to power the present monitor. A fast recharging circuit has been developed for the present application. The battery is initially charged at a constant high current (at approximately 3.5 C rate). The cell terminal voltage is continuously monitored. The charging current is switched automatically to the 0.1 C rate when the cell voltage reaches 1.85 V. It was found that a spent cell can be recharged to 90% capacity in less than 30 minutes.

To operate the unit, the operator performs the following simple tasks:

(1) Remove the unit from protective casing, strap loosely on the subject, and turn on power.

(2) Position unit to obtain a good pickup and then secure unit by tightening belt.

(3) If only a quick measurement of the heart rate is desired, hold monitor firmly on chest momentarily. If the "HOLD FOR INST." switch located on the side is depressed, a heart rate reading can be obtained in one or two seconds.

A prototype heart rate monitor having essentially the same features has been developed. No attempt has been made to miniaturize or optimize the various developmental circuitry. The functional block diagram for the monitor is shown in Figure 13.

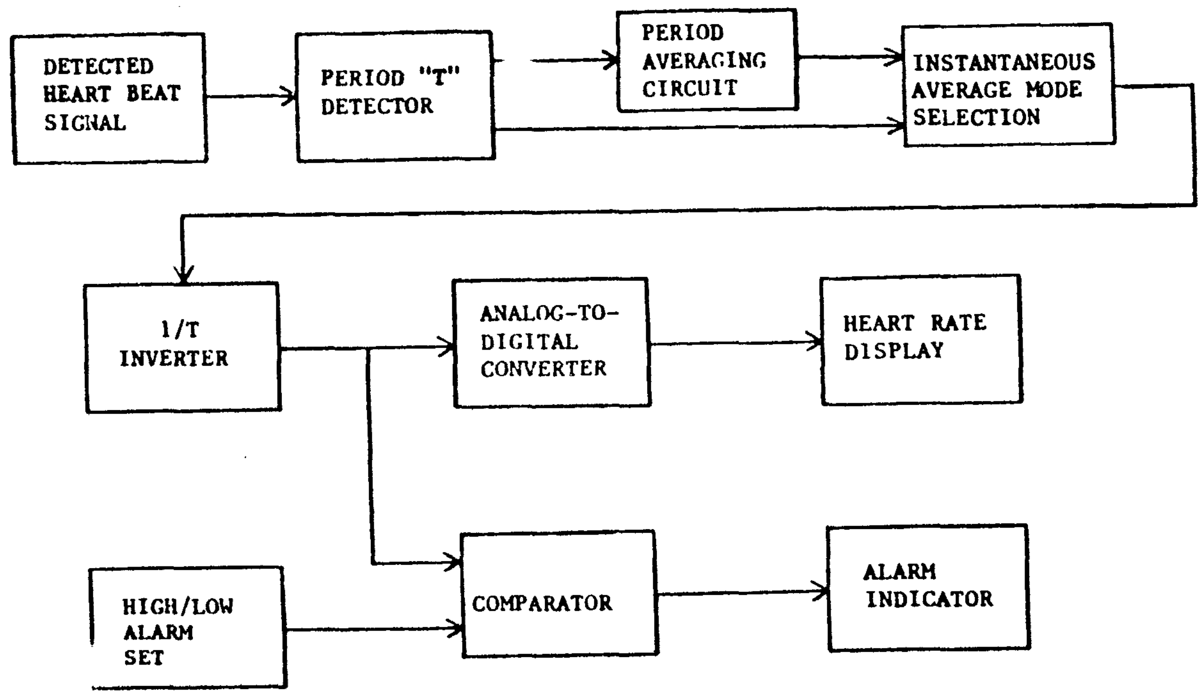


FIGURE 13 FUNCTIONAL BLOCK DIAGRAM OF HEART RATE MONITOR

3.0 PERFORMANCE

Prototype heart rate monitors based on Doppler effect and cardiac wave pickup have been constructed. The accuracy in heart rate measurement was found to be within 1% (or $\pm$ 1 BPM, whichever is greater) when calibrated against a crystal-controlled standard. The performance and field test results are presented in this section.

3.1 Ultrasound Heart Rate Monitor

The ultrasound monitor has been field-tested on board an Army M113 tracked vehicle. During idling, regular heart rhythm and rate measurements were obtained. However, when the vehicle was in motion on a paved track, the monitor failed to detect the heart beat. A recording of the beat frequency output from the ultrasound receiver stage is shown in Figure 14. The data were obtained with the test subject on a stretcher located at the right (rear) side of the vehicle. The subject wore a regular shirt plus the CW protective jacket. The transmitter-receiver pickup array, which is not mounted to the bottom of the monitor at present stage of development, was attached to the chest by means of an elastic band.

Field test data indicated that the excessive vibration encountered is at such a level that the signals were completely overwhelmed and buried by noise. The contributing factors to noise generation are as follows:

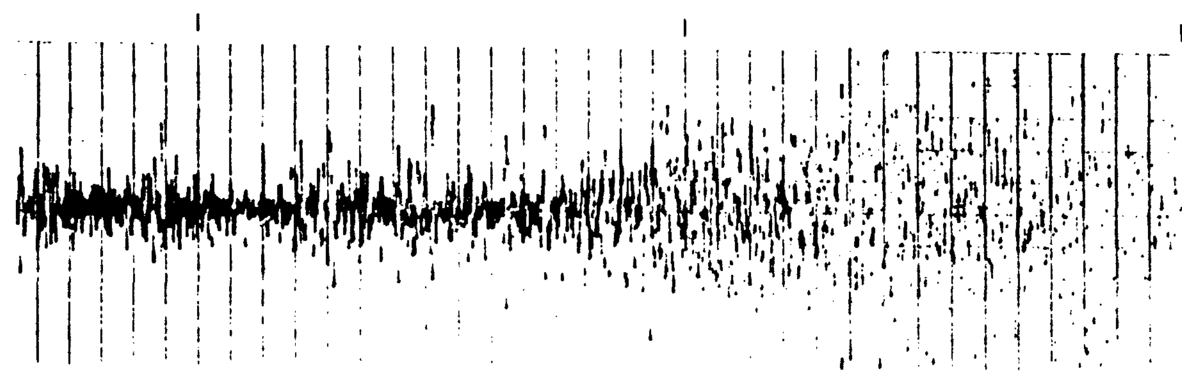
(1) Vehicle motion may have induced a certain degree of movement of the cardiac mass.

(2) Due to the difference in mass density between body tissue and the monitoring device, relative motion and velocity induced by vibration is unavoidable.

(3) Vibration causes random decoupling of the transmitter and receiver creating transient signals.

(4) In conventional Doppler velocity measurements, the beat frequency depends only on the relative velocity. However, in the present application, this is no longer true.

(a) VEHICLE IDLING



(b) VEHICLE IN MOTION

FIGURE 14 RESPONSE OF BEAT FREQUENCY DURING IDLING AND IN SLOW MOTION ON PAVED TRACK

If the transmitter-receiver is motionless, the frequency shift is governed by the following equation

$$\Delta f_h = 2f \frac{V_h}{C_b}$$

where f is the ultrasound frequency, V_h, the heart velocity and C_b, the speed of sound pertaining to the body.

If the transmitter-receiver moves independently of the heart (i.e., assuming the heart is not moving), the frequency shift is now given by the following equation:

$$\Delta f_{tr} = 2f \frac{V_{tr}}{C_{cw}}$$

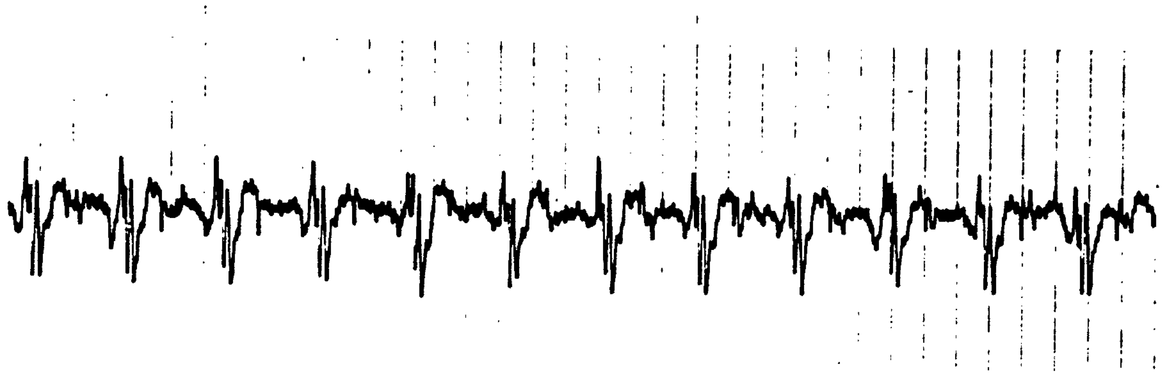
where the subscript tr and cw pertain to transmitter-receiver, and CW garment material, respectively.

Because the speed of sound of the body tissue is expected to be much larger than that of the CW protective garment material, the Doppler shift due to vibration movement of the transmitter-receiver is much greater. In other words, the noise signal would be weighted greater by the factor C_b/C_{cw} . The speed of sound for body tissue is approximately 1550 m/sec, and the sound speed in CW garment material is still not known. The speed of sound for sheepskin is 470 m/sec (Ref. 2).

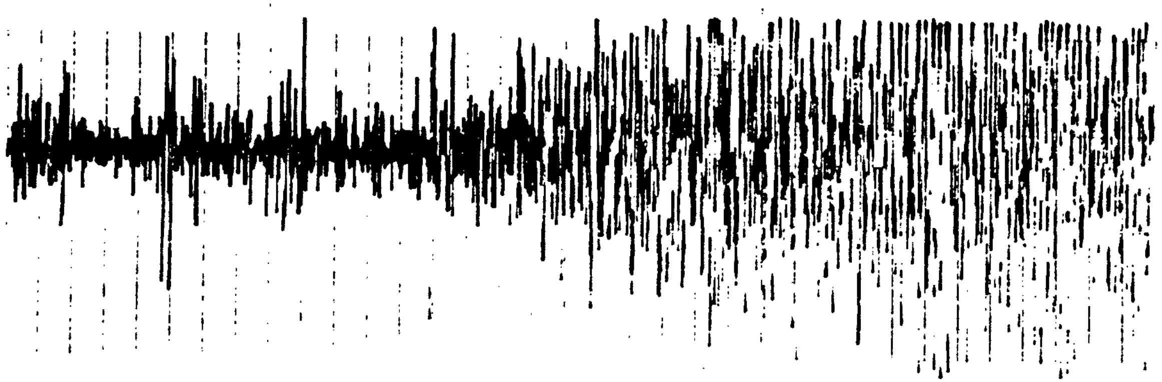
The above effect can also be explained in terms of wave reflection. In operation, ultrasound is constantly reflected from any interface whenever there is an abrupt change in the acoustic impedance such as the chest wall or the surface layer of the CW garment. These reflected waves normally do not cause any problem because their frequency is identical to the incident wave. If the transmitter-receiver array is moving relative to the skin, the reflected wave will contribute to the frequency shift signal.

3.2 Cardiac-Wave Heart Rate Monitor

The performance of the cardiac wave monitor appears to be superior to the Doppler monitor in terms of noise immunity. However, this monitor also failed to provide a usable heart rate measurement when the vehicle is in motion. A recording of the unprocessed cardiac wave signal is given in Figure 15.



(a) ON GROUND



(b) VEHICLE IN MOTION

FIGURE 15 RESPONSE OF HYDROSTATIC WAVE ON GROUND AND WHEN VEHICLE IS IN MOTION

4.0 CONCLUSIONS AND RECOMMENDATIONS

The detection of heart beat by the ultrasound Doppler technique and by direct cardiac shock wave recording have been investigated. Prototype heart rate monitors based on these two methods have been developed, constructed and field tested. Relevant data, design considerations and underlying principles have been presented. The goal of obtaining accurate heart rate measurements noninvasively, without compromising the integrity of the chemical warfare protective garment and body wrap, has largely been achieved. However, these monitors failed to provide proper measurement of the heart rate on board a moving M113, tracked vehicle. The important attributes of the heart rate monitors are summarized as follows:

(1) Both prototype heart rate monitors provided accurate measurements of the heart rate except when used on board a moving M113 vehicle.

(2) Placement and positioning of these monitors on the chest is not critical.

(3) The elastic strip used to secure the monitor on the chest did not exert undue pressure on the subject.

(4) Heart rate can be measured even when the subject wears undershirt and sweater in addition to the CW garment.

(5) The monitors can be operated in a handheld fashion to obtain a quick measurement. A heart rate reading can be obtained within one or two seconds in the "INSTANT UPDATE" mode.

(6) The monitor based on cardiac wave detection is in general, superior to the ultrasound monitor. Because it is a passive device which does not require an energetic beam or wave to interact with the body, it is smaller, lighter, less complicated, easier to maintain, and requires less battery capacity. The size after miniaturization is estimated to be approximately 7 cm x 12 cm x 3 cm.

(7) With regard to operation in the tracked field ambulance, it appears certain that in addition to the movement of the heart, the relative motion between the monitor and the body needs to be monitored separately in order to cancel noise artefacts from the composite signal. In the case of Doppler monitor, high frequency waves or overtones can be utilized to monitor the vibration of the heart rate monitor relative to the body. Such high frequency ultrasound does not penetrate deep into the body and can be designed to reflect completely off the skin surface.

Based on test results, discussions and observations presented in previous sections, the following recommendations are ventured:

(1) Complete Phase II development of the heart rate monitor. Despite the shortcoming of not being able to properly function on the M113 vehicle, the monitor is still quite useful for monitoring and for quick assessment of the heart condition especially in situations where the casualty is in the body wrap and application of a blood pressure cuff is precluded.

(2) Employ cardiac wave detection method in the design of the heart rate monitor.

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